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The force output of handle and pedal in different bicycle-riding postures

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The purpose of this study was to analyse the force output of handle and pedal as well as the electromyography (EMG) of lower extremity in different cycling postures. Bilateral pedalling asymmetry indices of force and EMG were also determined in this study. Twelve healthy cyclists were recruited for this study and tested for force output and EMG during steady state cycling adopting different pedalling and handle bar postures. The standing posture increased the maximal stepping torque (posture 1: 204.2 ± 47.0 Nm; posture 2: 212.5 ± 46.1 Nm; posture 3: 561.5 ± 143.0 Nm; posture 4: 585.5 ± 139.1 Nm), stepping work (posture 1: 655.2 ± 134.6 Nm; posture 2: 673.2 ± 116.3 Nm; posture 3:

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1852.3 ± 394.4 Nm; posture 4: 1911.3 ± 432.9 Nm), and handle force (posture 1: 16.6 ± 3.6 N; posture 2: 16.4 ± 3.6 N; posture 3: 26.5 ± 8.2 N; posture 4: 41.4 ± 11.1 N), as well as muscle activation (posture 1: 13.6–25.1%; posture 2: 13.0–23.9%; posture 3: 23.6–61.8%; posture 4: 22.5–65.8%) in the erector spine, rectus femoris, tibialis anterior, and soleus. However, neither a sitting nor a standing riding posture affected the hamstring. The riding asymmetry was detected between the right and left legs only in sitting conditions. When a cyclist changes posture from sitting to standing, the upper and lower extremities are forced to produce more force output because of the shift in body weight. These findings suggest that cyclists can switch between sitting and standing postures during competition to increase cycling efficiency in different situations. Furthermore, coaches and trainers can modify sitting and standing durations to moderate cycling intensity, without concerning unbalanced muscle development.

**KEYWORDS** biomechanics, exercise, kinesiology

**INTRODUCTION**

Bicycle-riding postures affect patterns of movement in the lower extremities, and different postures require different muscles activation (Bini, Hume, & Croft, 2011) and alter pedalling efficiency (Peveler & Green, 2011). Recent studies have focused on parameters in the cyclist’s sitting posture while riding, but cyclists also use a standing posture depending on road conditions. The standing posture is used when riding uphill, which increases energy consumption, changes the body's centre of gravity, and increases the range of motion of the ankle joint (Alvarez & Vinyolas, 1996; Caldwell, Li, McCo, & Hagberg, 1998; Soden & Adeyefa, 1979) to enhance the maximal stepping torque output (Alvarez & Vinyolas, 1996; Caldwell et al., 1998). Studies have revealed that using a standing riding posture elevates the trunk and prevents low back pain (Duc, Bertucci, Pernin, & Grappe, 2008). Other studies have indicated that regardless of road conditions, whether a cyclist adopted a standing or sitting riding posture does not affect physiological indices, such as riding economy, pedalling efficiency, maximal heart rate, and rate of perceived exertion (Harnish, King, & Swensen, 2007). However, the difference between standing and sitting riding postures may affect performance. In addition to seat position, handle positions alter the trunk angle during riding, affecting muscle activation in the lower extremities (Chapman et al., 2008; Savelberg, Van de Port, & Willems, 2003). Therefore, both handle position and riding posture should be studied together to determine more efficient application for cycling.
Previous studies have revealed that the difference between right and left pedalling torques ranges from 0.5% to 17% in sitting posture (Carpes, Rossato, Faria, & Bolli Mota, 2007; Carpes et al., 2008b), and that the power output of the human body is asymmetrical (Carpes, Bini, Mota, & Carpes, 2008a). Riding bicycle with a standing posture resulted in greater muscle activation and power output (Hansen & Waldeland, 2008; Li & Caldwell, 1998); however, riding bicycle with a standing posture also exhibited a marked lateral asymmetry, which affected pedalling performance (Carpes et al., 2008b; Duc et al., 2008). Pedalling performance is reduced if pedalling asymmetry affects the rider's pedalling workload (Carpes, Mota, & Faria, 2010). Another study indicated that pedalling asymmetry might contribute to knee injuries due to overuse among both recreational and competitive cyclists (Holmes, Pruitt, & Whalen, 1994).

This study analysed the kinetic and electromyography (EMG) differences between sitting and standing cycling postures while observing pedalling asymmetry. The hypotheses were: (1) Stepping torque and levels of muscle activation in a standing posture are considerably higher than in a sitting posture; (2) The handle force during a standing riding posture is considerably higher than during a sitting posture; and (3) The stepping torque and muscle activation of the left and right sides are asymmetry.

METHODS

Participants

Twelve healthy cyclists (male: 9, female: 3) with left dominant leg (Age: 23.92 ± 1.56 years, height: 173.33 ± 5.60 cm, weight: 69.04 ± 11.86 kg) were recruited for this study. None of the participants had experienced lower extremity injuries that were serious enough to require medical treatment in the year before the study was conducted. The study was approved by the Medical Research Ethics Committee of local medical university hospital and all participants completed a statement of informed consent.

Bike Setting

A stationary flywheel bicycle (Magtonic Inc., Taiwan) was used in this study. The bicycle was equipped with: (1) a 3-force-component transducer (LFX-A-3KN, Kyowa Co., Japan) mounted under the handlebar to detect resultant force, representing the total amount of force applied to the handle bar; (2) a photoelectric sensor to determine the crank angle and record pedal positions; and (3) two torque sensors to measure right- and left-pedal torque. All signals were passed through a main amplifier and were acquired using an analogue-to-digital converter system (Biopac Systems Inc., USA) at a sampling rate of 1000 Hz.
Electromyography

To investigate muscle activity in the back and lower extremities, EMG signals obtained from muscle groups on both sides of the body were recorded from the erector spine (ES; back), rectus femoris (RF; thigh), hamstring (BF; thigh), tibialis anterior (TA; shank), and soleus (shank). The placements of the electrodes are described as follows: (1) In the ES, the electrode was placed at approximately the L4 level, in the midpoint between the lateral palpable border of the ES and the posterior superior iliac spine; (2) In the RF, the electrode was placed on the midpoint between the anterior superior iliac spine and the superior aspect of the patella; (3) In the BF, the electrode was placed on the midpoint between the distal ischial tuberosity and the popliteal fossa; (4) In the TA, the electrode was placed on the upper-third of the muscle, which extends from the tibial head to the medial malleolus; and (5) In the soleus (SOL), the electrode was placed on the midline of the calf muscle and close to the junction between the gastrocnemius and the Achilles tendon.

Active electrodes (TSD150 series, Biopac Systems Inc., Goleta, CA, USA) were used to record muscle EMG activities. The reference electrode was placed on the lateral malleolus of the left ankle. The electrodes (5 mm in diameter) were positioned with an interelectrode distance of 20 mm. Skin preparation before application of surface electrodes ensured that the interelectrode resistance was below 5 kΩ. The EMG signal sampling rate was set at 1000 Hz (preamplifier: common mode rejection ratio = 95 dB; impedance = 100 MΩ; gain = 350). All EMG signals were recorded using an acquisition system (Biopac MP150, Biopac Systems Inc., Goleta, CA, USA) into a personal computer. Before the start of the cycling trials, the electrodes were adhered to participants’ muscles to record maximal voluntary isometric muscle contraction signals, which were used as the normalized standard signals.

Protocol

Four common riding postures (Duc et al., 2008; Slane, Timmerman, Ploeg, & Thelen, 2011) were tested. As shown in Figure 1, these postures were: (1) hands on the midpoint of the handlebar in a sitting posture, (2) hands on the sides of the handlebar in a sitting posture, (3) hands on the sides of the handlebar in a standing posture, and (4) hands on the front of the handlebar in a standing posture. Each participant performed each of the four riding trials within 2 min, with a 2-min break between trials to prevent muscle fatigue. Metronomes were used to control the pedalling cadence at 50 rpm (Hansen, Andersen, Nielsen, & Sjogaard, 2002). When the participants’ riding patterns became stable and consistent (typically 4–5 cycles after the start of the trial), we marked the signals and recorded the following data for 1 min: (1) hand force applied to the handlebar, (2) stepping torque applied to the crank axis, and (3) EMG activity in the muscles of the back and lower extremities.
Data Analysis

To compare the performance of the right and left legs, we defined the period of leg activity according to the crank angle. The crank angle at 0° was equal to top dead point of the right pedal, 0–180° was defined as the right-leg period, and 180–360° was defined as the left-leg period. Defining these periods enabled calculations to compare the torque values and EMG signals of both legs.

The torque strain gauge was used to measure the torque applied by participants from the crank axis. The maximal stepping torque was measured and the stepping work was calculated by multiplying the torque and the crank angle. The force data (maximal stepping torque, stepping work, and handle force) were used to determine the bilateral asymmetry index. The
asymmetry index (AI) was calculated by the percentage difference between right (R) and left (L) sides (Equation 1) (Karamanidis, Arampatzis, & Bruggemann, 2003; Knapik, Bauman, Jones, Harris, & Vaughan, 1991). The reference value of 10% was considered asymmetric by other studies (Carpes et al., 2007, 2010).

\[
AI(\%) = \left[ \frac{(R - L)}{\frac{1}{2}(R + L)} \right] \times 100\%.
\]  

(1)

In the EMG analysis, same as in the stepping torque and handle force analyses, the cycle was divided into right- and left-leg periods. EMG signals were filtered using a high-pass filter at 25 Hz and a low-pass filter at 500 Hz. Then the filtered EMG signals were calculated using the root-mean-square method, and the moving window was set at 50 samples (rEMG). Finally, the rEMG was divided by the maximal value of rEMG obtained from the isometric maximal voluntary muscle contraction values to acquire the normalized EMG.

Statistical Analysis
A two-way analysis of variance (ANOVA) with repeated measures was used to analyse all the data (4 postures × 2 sides). Tukey’s post hoc tests were used to further identify the statistically significant mean differences. The level of significance was set at \( \alpha = .05 \). Partial eta squared (partial \( \eta^2 \)) and observed power (\( \beta \)) values were calculated to complete the analysis. Partial \( \eta^2 \) was used to calculate effect sizes, with outputs of 0.5 or greater considered a large effect size, 0.1–0.5 a moderate effect size, and less than 0.1 a small effect size (Field, 2009). The descriptive statistics (mean value ± standard deviation) was used to analyse AI.

RESULTS
The two-way ANOVA revealed that no significant interaction were found for the maximal stepping torque \( (F_{3,66} = .103, p = .958, \text{partial } \eta^2 = .005, \text{ power} = .068) \) stepping work \( (F_{3,66} = .005, p = .999, \text{partial } \eta^2 = .015, \text{ power} = .051) \) and handle force \( (F_{3,66} = .020, p = .996, \text{partial } \eta^2 = .001, \text{ power} = .053) \). However, the main effect for different postures on the maximal stepping torque \( (F_{3,66} = 251.645, p = .000, \text{partial } \eta^2 = .920, \text{ power} = .999) \), stepping work \( (F_{3,66} = 212.489, p = .000, \text{partial } \eta^2 = .906, \text{ power} = .999) \) and handle force \( (F_{3,66} = 160.286, p = .000, \text{partial } \eta^2 = .879, \text{ power} = .999) \) were significant. Post hoc comparisons showed that postures 1 and 2 (sitting posture) were significantly lower than postures 3 and 4 (standing posture), posture 3 was significantly lower than posture 4 on the maximal stepping torque (posture 1: 204.2 ± 47.0 Nm; posture 2: 212.5 ± 46.1 Nm; posture 3: 561.5 ± 143.0 Nm; posture 4: 585.5 ± 139.0 Nm), stepping work
(posture 1: 655.2 ± 134.6 Nm; posture 2: 673.2 ± 116.3 Nm; posture 3: 1852.3 ± 394.4 Nm; posture 4: 1911.3 ± 432.9 Nm), and handle force (posture 1: 88.9 ± 20.1 N; posture 2: 90.6 ± 20.0 N; posture 3: 130.3 ± 41.8 N; posture 4: 213.7 ± 61.0 N). The right and left sides do not show any significant difference in maximal stepping torque, stepping work, and handle force (Figure 2). But the asymmetry indices of maximal stepping torque (posture 1: 22.0 ± 14.4%; posture 2: 17.9 ± 9.9%; posture 3: 6.4 ± 4.8%; posture 4: 5.1 ± 4.0%) and stepping work (posture 1:

Figure 2. The results of maximal stepping torque, stepping work and maximal handle force in four riding postures. *Significant main effect of posture conditions (P < .05); *Significant difference between two postures (P < .05).
20.2 ± 12.1%; posture 2: 18.3 ± 7.9%; posture 3: 7.3 ± 4.8%; posture 4: 7.8 ± 3.9%) of sitting posture exceed 10% which was considered asymmetric.

The two-way ANOVA revealed that no significant interaction were found for the TA ($F_{(3,66)} = 2.531, p = .065$, partial $\eta^2 = .103$, power = .600), soleus ($F_{(3,66)} = .899, p = .446$, partial $\eta^2 = .039$, power = .237), quad ($F_{(3,66)} = 1.575, p = .204$, partial $\eta^2 = .067$, power = .396), ham ($F_{(3,66)} = .743, p = .530$, partial $\eta^2 = .033$, power = .200), and back muscles ($F_{(3,66)} = .492, p = .689$, partial $\eta^2 = .022$, $\beta = .145$). However, the main effect for different postures on the TA ($F_{(3,66)} = 32.455, p = .000$, partial $\eta^2 = .596$, power = .999), soleus ($F_{(3,66)} = 101.524, p = .000$, partial $\eta^2 = .822$, power = .999), quad ($F_{(3,66)} = 92.660, p = .000$, partial $\eta^2 = .808$, power = .999), ham ($F_{(3,66)} = .679, p = .568$, partial $\eta^2 = .030$, power = .186), and back muscles ($F_{(3,66)} = 53.131, p = .000$, partial $\eta^2 = .707$, power = .145) were significant. Post hoc comparisons showed that postures 1 and 2 (sitting posture) were significantly lower than postures 3 and 4 (standing posture) on TA (posture 1: 25.1 ± 12.0%; posture 2: 23.6 ± 11.8%; posture 3: 52.2 ± 23.0%; posture 4: 50.4 ± 21.5%), soleus (posture 1: 24.6 ± 13.1%; posture 2: 23.9 ± 11.7%; posture 3: 61.8 ± 12.2%; posture 4: 65.8 ± 16.7%), quad (posture 1: 13.6 ± 8.4%; posture 2: 13.0 ± 8.6%; posture 3: 57.7 ± 22.7%; posture 4: 49.9 ± 17.7%), and back muscles (posture 1: 16.6 ± 11.2%; posture 2: 15.3 ± 7.9%; posture 3: 35.7 ± 13.4%; posture 4: 37.0 ± 15.2%). However, the ham (posture 1: 16.4 ± 11.1%; posture 2: 21.9 ± 17.1%; posture 3: 23.6 ± 21.7%; posture 4: 22.5 ± 17.5%) showed no significant difference among the four riding postures. On the other hand, the right and left sides do not show any significant difference in muscle activation (Figure 3).

![Figure 3](image_url)

Figure 3. The comparison of the normalized electromyography signal of each muscle in sitting and standing postures. *Significant difference between sitting and standing postures ($P < .05$).
DISCUSSION

In accordance with our hypothesis, the findings of this study demonstrated that the standing posture increased not only the maximal stepping torque and stepping work, but also handle force, as well as muscle activation in the ES, RF, TA, and SOL. However, using a sitting or standing riding posture did not affect the BF. Besides, no significant difference was detected between the right and left legs in any condition. On the other hand, changing handle bar position did not affect handle bar force, maximal stepping torque, and stepping work in sitting riding posture. However, different handle bar positions showed significantly different handle bar force, maximal stepping torque, and stepping work in standing riding posture.

When using the sitting riding posture, most of the cyclist’s body weight is supported by the bicycle seat. Using the standing riding posture, the weight supported by the bicycle seat is transferred to the upper and lower extremities. Given the same resistance, the standing riding posture was revealed to provide a higher stepping power output than the sitting posture (Hansen & Waldeland, 2008). Therefore, body weight is responsible for the higher stepping torque. Body weight was observed to be used more actively when adopting the standing rather than sitting cycling position. This is because the cyclist’s hips are further forward providing leverage over the crank arm, which a sitting posture does not provide (Fleming et al., 1998). When participants moved from the sitting posture to the standing posture, shifting of their weight from the seat to the handlebars and pedals accounted for the change in handle force. The handle force results show that the maximal handle force in the standing riding posture is approximately 1.5 times that of the sitting riding posture. This is because part of body weight, originally supported by the bicycle seat, must be supported by the upper extremities when the riding posture changes from sitting to standing. The increased participation of the upper extremities can also be used as a training approach for upper extremities. Cyclists often switch between sitting and standing postures during uphill cycling. This switching increases cycling efficiency in real conditions. In addition, this study also found that change handle position could affect the force output on the upper and lower limbs only with standing posture. It has been reported that cycling in standing posture, the arms pull up and back during the power stroke, and push down and forward during the upstroke (Stone & Hull, 1993).

The normalized EMG amplitude of all muscle groups considerably increased when the cyclist changed his riding posture from sitting to standing because more muscles must be recruited to support body weight originally supported by the bicycle seat (Miller, Peach, & Keller, 2001) with the exception of the hamstring muscle. Standing up from the bicycle seat causes a change in the muscle recruitment pattern: body weight is no longer supported by the bicycle seat, and the activity of the lower extremities muscle groups increases to stabilize the joints. Brown et al., noted that muscle activation patterns in the ankle muscles change considerably to maintain a relatively consistent position of the ankle even.
late in the upstroke, and both net dorsiflexor ankle torque and TA activity increases as body orientation becomes more vertical (Brown, Kautz, & Dairaghi, 1996). Our results regarding ankle muscle groups were consistent with this. The left and right soleus and TA revealed higher normalized EMGs in the standing posture than in the sitting posture. In addition, higher muscle activation of the quadriceps was considered to control knee-joint balance in the standing posture. The normalized EMG amplitude of hamstring muscles might not have increased because of the inertial effect of flywheel in the period from the bottom dead point to the top dead point. Cyclists do not need to actively flex their knees when riding a flywheel bicycle during this period because of the existing angular momentum of the flywheel. The erector spine muscle exhibited a higher EMG activation in the standing posture than in the sitting posture. Muscles in the trunk and arms provide a counterbalancing force to the lower extremities during pedalling. The hands, arms, shoulders, abdomen, and back form a muscular sling, which rhythmically moves back and forth in supporting the trunk and pelvis (Schmidt, 1998). This result can be applied to the design of training programs or rehabilitative programs using bicycles.

Indoor cycling has been recognized as a recreational and sporting activity that has many therapeutic qualities. Fleming et al. suggested that stationary cycling at the appropriate cadence and resistance is an effective rehabilitation exercise for patients with anterior cruciate ligament injury as it can increase muscle activity without subjecting the ligament to undue strain (Fleming et al., 1998). With an understanding of the normal muscle recruitment pattern of this exercise, sports physiotherapists can focus on a particular phase of the cycling action to train particular muscle groups. Previous studies have demonstrated no difference in submaximal uphill cycling performance between the seated and standing positions. During maximal cycling performance, however, standing position could be more energetically effective than sitting (Hansen & Waldeland, 2008). And our findings confirm the findings of these studies.

Riding asymmetry has been observed to be a factor causing knee injuries from overuse (Holmes et al., 1994). In this study the EMG amplitude results indicated that no major differences existed between the right and left side muscle groups in the sitting or standing riding postures. However, the asymmetry indices of maximal stepping torque and stepping work of sitting posture exceed 10% which is considered asymmetric. The stepping torque on the left side (dominant leg) tended to be greater than on the right side. In the sitting posture, 10 of the 12 participants exhibited greater stepping torque on the left side than on the right side, even after changing the position of their hands on the handlebar. In the standing posture, although participants used weight shifting to accelerate the flywheel and showed inconsistent stepping torque results, the average stepping torque values still indicate that torque on the left side was greater than on the right side. A more even distribution of the forces between two legs would reduce the knee loads transmitted by the dominant leg and therefore reduce overuse injuries.
CONCLUSION

As expected, when a cyclist changes his posture from sitting to standing, the upper and lower extremities are forced to produce more force output because of the shift in body weight, which was originally supported by the bicycle seat in the sitting posture. Consequently, the cyclist can always change the grip position during sitting which does not affect the power output and may also reduce the pressure on the hand. However, riding on a stationary bike with sitting posture showed higher riding asymmetry. These findings suggest that cyclists can switch between sitting and standing postures during competition to increase cycling efficiency in different situations. Furthermore, coaches and trainers can modify sitting and standing durations to moderate cycling intensity, without concerning unbalanced muscle development. This result can also be applied to the design of training programs or rehabilitative programs using stationary bike.

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DISCLOSURE STATEMENT

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